AN AUTOMATIC CALIBRATION METHOD FOR BIPLANAR X-RAY 3-D RECONSTRUCTION

H. El Dakhakhni[1], M.R. Mahfouz[1], E.E. Abdel Fatah [1], E. Pritchard[1]
[1]Biomedical Engineering Department, University of Tennessee, Knoxville, USA
e-mail: mmahfouz@utk.edu

Abstract– **The reconstruction of three-dimensional models from two biplanar radiographs has been a topic of research to reduce patient radiation dose and preoperative costs for creating models from Computed Tomography (CT). In this context, we introduce a new approach for biplanar three-dimensional (3-D) model reconstruction, based on recreation of the X-ray 3-D scene. This approach is expected to decrease the time required to build 3-D models. We also introduce an automatic method for the calibration of the radiological environment of the femoral radiographs based on a calibration cube phantom. Image processing techniques used to extract the projected bead locations will be described, as well as the methods for utilizing these locations to calibrate the radiological environment.**
***Keywords*- Calibration, Biplanar, 3-D Reconstruction**

I. INTRODUCTION

Due to advances in computational and storage technology, three-dimensional – and higher dimensionality - applications such Computer Aided Surgery (CAS), quantitative 3-D measurements and pathological deformation estimation [1] have become affordable. Such applications require patient-specific 3-D models of the organ(s) of interest. Traditionally, such models are created by introducing the patient to an imaging modality that performs sequential, high resolution cross-sectional scanning of the area enclosing the organ of interest. Next, 3-D reconstruction methods are implemented to integrate the captured 2-D images into a 3-D volume. The volume is then segmented to extract the organ of interest from surrounding tissue, allowing a surface model of the organ to be formed.

Although Computed Tomography (CT) is usually the gold standard for 3-D bone studies, effective radiation dose delivered to the patient in such studies far exceeds that of traditional radiography. For example, a chest X-ray's effective dose would range from 0.01 to 0.02 mSv, while performing a chest CT would increase the does from 5 to 7 mSv [2].

Several attempts have been made to estimate bone models using only two radiographic projections. For example, Sato et al. [3] projected a femoral model to biplanar images using the system projection matrix. Using a 2-D image fitting algorithm, this projection is fitted to each of the radiographic images, then back-projected using the inverse projection matrix to create an estimate of the actual model. Although such a method could have sufficient results for certain requirements, the lack of statistical insight into feasible deformations of the human femur could yield unrealistic results. Benameur et al. [1] used the Stochastic Expectation/Maximization optimization algorithm [4] to optimize a Probabilistic PCA model [5] of the rib cage. The optimization function was based on minimizing a logarithmic energy function between the projection of the candidate model and the projection of the edge detected ribs. In [6], an anatomic atlas of the distal femur was deformed using a Kriging algorithm [7] to minimize the error between defined 2-D radiograph contours and the projection of their corresponding atlas 3-D contours.

We suggest an alternative approach for estimating the patient-specific femoral model. First, models of a statistically representative population of femurs can be created by segmentation of 3-D CT volumes of the hip. These models would be reduced using Principle Component Analysis (PCA) and further studied to model the variation among them to create a Probabilistic PCA (PPCA) model of the femur. A mean instance of the model will then be created and inserted into a 3-D scene that recreates the radiological environment of the X-ray machine for both images (Fig.1). Projections of the mean model would be taken and compared to the actual X-ray images. According to an energy function, the model would be allowed to deform by changing its principle components in an iterative optimization process until its projection produces an acceptable fit. The paper is organized as follows: Section II presents the methodology used for bead extraction, bead identification, and X-ray source estimation, Section III presents experimental results, Section IV provides corresponding discussion, and Section V concludes.

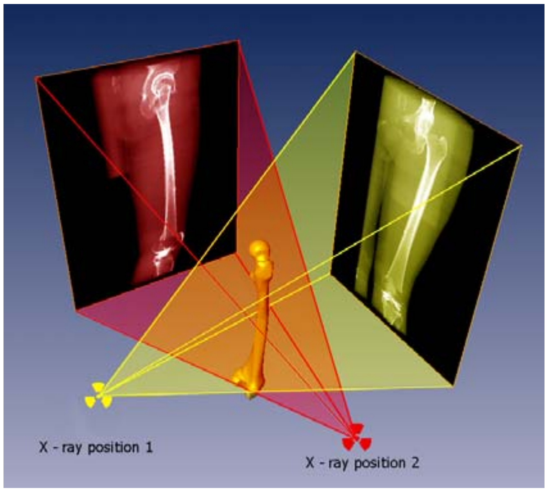


Fig. 1 Recreation of X-ray 3-D scene

II. METHODOLOGY

A 90x90x90 mm plastic cube with 15 mm displaced steel beads forming a 5x5x5 grid was used as our calibration phantom (Fig. 2). This phantom was securely attached to the mid-shaft of five dry femurs and for each femur, an anterior-posterior and a lateral X-ray was taken. The proximal end of the femur was raised using radio-transparent foam so that the femoral anatomical axis would be parallel to the image plane, simulating life situations. The resultant radiographs were digitized using a Vidar® Sierra Plus film digitizer at 300 dpi (Fig. 3).

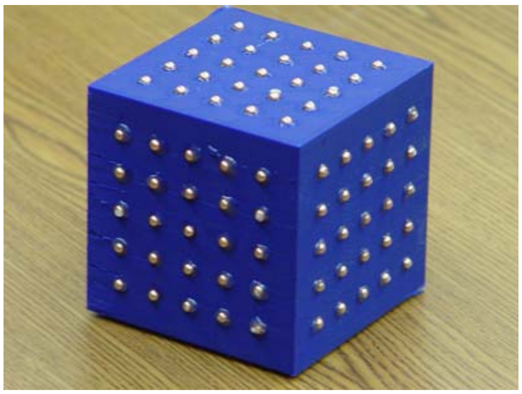
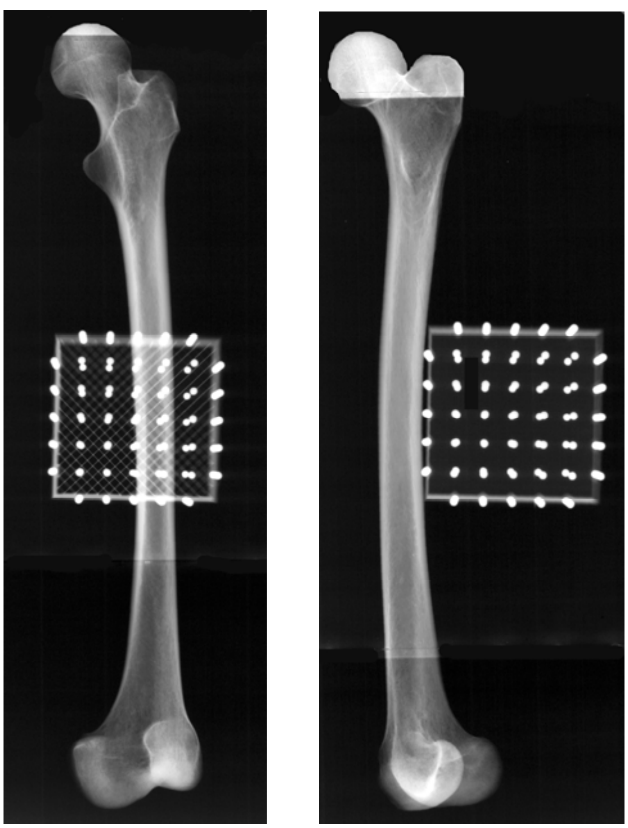


Fig.2 Calibration cube



(a) (b)

Fig. 3 (a) AP View (b) Lateral View

A. Bead Extraction

Our algorithm of extracting the projections of the steel beads from the digitized radiographs depends mainly on morphological operators. Such operators expect a well defined structuring element as an input parameter. Therefore, it is necessary that the radius of the beads' projections be well known before attempting to use a disk structuring element to extract them. Factors affecting this radius include the relative transformations between the X-ray source, the calibration cube and the image plane, and the resolution of the digitized image. In order to automatically detect the radius of the beads projections, we detected the number of separate structures in the image as the morphological opening disk size changes. The optimal disk size was found by detecting the size at which the number of structures starts to decrease significantly (Fig. 4). Knowing the size of the bead projections, we were able to remove the femur from the image. Any structure whose area is larger than the maximum possible area of connected bead projections, or of width or height greater than the diameter of the maximum possible connected beads was removed from our image. Any other persistent non-bead structure was removed by spatial location inspection relative to the existing structures' centroid.

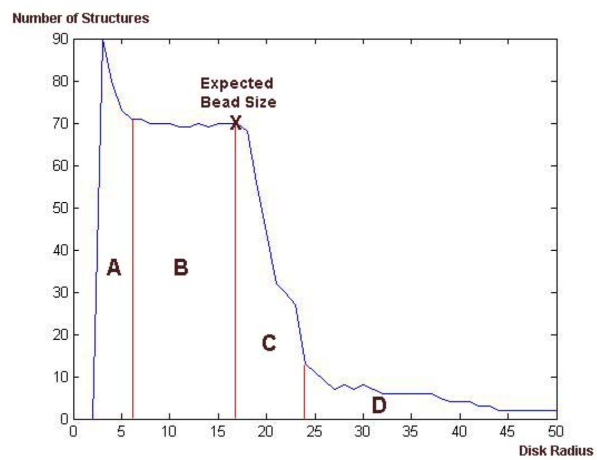


Fig. 4 (A) Small femoral structures of high intensity are responsible for a high structure count. (B) Beads dominate the structure count (C) As the disk size exceeds the bead diameter, the structures' count fall sharply due to the loss of bead projections. (D) All beads eventually dissappear, leaving large structures that belong to the femur.

B. Bead Identification

Having extracted all the beads projections from the image, the next step is to identify the bead corresponding to every bead projection. We used a bucketing algorithm [8] to identify the X and Y coordinates of the "bucket" of each structure. Every bucket is then further manipulated to extract the center of each of the two possible bead projections located inside it. A Distance Transform

algorithm [9] was used to separate connected projections. The Z-coordinate was found by comparing distances between beads of collinear buckets, with the larger distance corresponding to the cube face nearer to the X-ray source (Fig. 5).

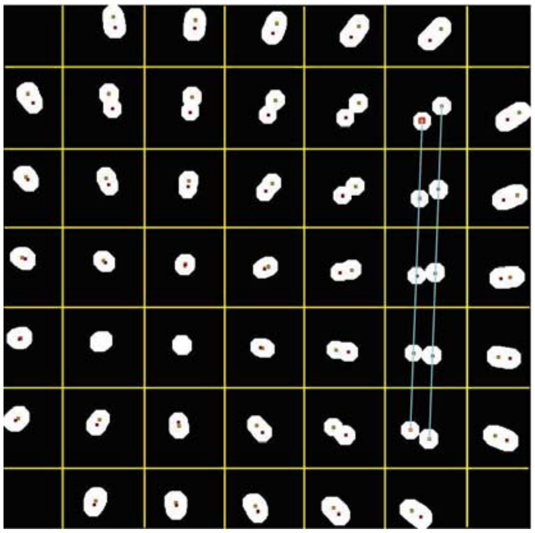


Fig. 5 Point correspondences bucketing showing the Z-Coordinate identification of the beads corresponding to the projections. Larger magnification clearly corresponds to the cube face closer to the X-ray source.

C. X-ray Source Localization

Using the correspondence between the location of the beads in the calibration coordinate frame and their 2-D projections in the image coordinate frame, the X-ray source location is estimated. We use a non-linear least squares method as described in [10]. Specifically, given an image point P, a 3-D model point Q, and the vector containing the correct pose parameters β, let the transformation to project Q onto the image be given by the function $f(\beta,Q)$ such that $P = f(\beta,Q)$. The vector function f represents a perspective projection (whose parameters are known from calibration). Linearizing around a current pose solution β, we get:

$$\Delta P = \left(\frac{\partial f}{\partial \beta}\right)\Delta\beta \qquad (1)$$

Given at least four point correspondences, we can solve for the correction term $\Delta\beta$. This process is iterated until the solution converges. The final solution is the pose β that minimizes the squared Euclidean distance between the observed image points and the projected model points on the image plane. A good initial guess for the cube pose can be obtained by assuming that the X and Y coordinates of the cube and X-ray source coordinate systems' coincide. Furthermore, assuming that the cube planes facing the X-ray source are parallel to the image planes, the Z-direction distance can be calculated by substituting in the radiography environment equations [11]:

$$\chi_1 = \frac{SID}{SID - OID_1} \qquad (2)$$

$$\chi_2 = \frac{SID}{SID - OID_2} \qquad (3)$$

$$OID_1 - OID_2 = \alpha \qquad (4)$$

where χ_1, χ_2 are the calculated magnification factors for the bead distances of the cube planes parallel to the image plane, SID is the X-ray source to image plane distance, OID_1, OID_2 are the distances between the parallel cube planes and the image plane, and α is the cube side width. By solving equations (2), (3), (4), a good estimate of the actual distance between the X-ray source and cube phantom locations can be obtained.

III. Results

We verified the resultant pose by re-projecting the phantom beads onto the image plane using the final calculated pose. This was applied on 10 digitized radiographs of five different dry femurs. The mean sum of squared error obtained was 0.172 mm and the maximum sum of squared error was 0.244 mm.

IV. Discussion

Anatomical statistical models are usually used in the 3-D reconstruction of patient-specific femoral models . Unfortunately, instances of such models are usually composed of a large number of vertices. Therefore, software computation required to project such a model is expected to be computationally expensive in an iterative optimization algorithm. Our attempt to use 3-D scene recreation carries the burden of vertex projection to hardware implemented software such as OpenGL. The output of this projection would be an off-screen rendered image that would be ready for comparison with the associated radiograph.

We discussed the methods used to automatically estimate the radiological environment setup parameters. In the bead projection location extraction process, it should be pointed out that the bucketing technique described used could fail if the angle between the image plane and the cube planes increases to the limit that causes interference between non-corresponding buckets. Other methods such as the interpretation search tree method [13] can be used for larger rotation angles. Furthermore, although only four point

correspondences can be sufficient to estimate the X-ray source location, increasing the number of correspondences was proved to decrease the error resulting from the least squares iterative technique [12]. The choice of the points should be well distributed throughout the cube for best results.

A final step would be required in order to use the calculated environment parameters. The relative transformation between the femur and its attached calibration phantom should be calculated. The femoral coordinate system should be defined according to anatomical landmarks that can be extracted automatically from both biplanar projections. After that, the transformation matrix relating the femoral and phantom cube coordinate systems can be used to calculate the pose of the femur in the X-ray environment.

V. CONCLUSION

This paper introduces a new approach for 3-D biplanar reconstruction which has a potential of decreasing the reconstruction time. A calibration method for such a system was presented and proven to have a high accuracy for small relative poses.

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